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DEMAND-BASED CARDIAC FUNCTION THERAPY

Field of the Invention

This patent application pertains to methods and apparatus for the treatment of cardiac disease. In particular, it relates to methods and apparatus for improving cardiac function with electro-stimulatory therapy.

Background

Implantable cardiac devices that provide electrical stimulation to selected chambers of the heart have been developed in order to treat a number of cardiac disorders. A pacemaker, for example, is a device which paces the heart with timed pacing pulses, most commonly for the treatment of bradycardia where the ventricular rate is too slow. Atrio-ventricular conduction defects (i.e., AV block) and sick sinus syndrome represent the most common causes of bradycardia for which permanent pacing may be indicated. If functioning properly, the pacemaker makes up for the heart's inability to pace itself at an appropriate rhythm in order to meet metabolic demand by enforcing a minimum heart rate. Implantable devices may also be used to treat cardiac rhythms that are too fast, with either anti-tachycardia pacing or the delivery of electrical shocks to terminate atrial or ventricular fibrillation.

Implantable devices have also been developed that affect the manner and degree to which the heart chambers contract during a cardiac cycle in order to promote the efficient pumping of blood. The heart pumps more effectively when the chambers contract in a coordinated manner, a result normally provided by the specialized conduction pathways in both the atria and the ventricles that enable the rapid conduction of excitation (i.e., depolarization) throughout the myocardium. These pathways conduct excitatory impulses from the sino-atrial node to the atrial myocardium, to the atrio-ventricular node, and thence to the ventricular myocardium to result in a coordinated contraction of both atria and both ventricles. This both synchronizes the contractions of the muscle fibers of each chamber and synchronizes the contraction of each atrium or ventricle with the contralateral atrium or ventricle.

Without the synchronization afforded by the normally functioning specialized conduction pathways, the heart's pumping efficiency is greatly diminished. Pathology of these conduction pathways and other inter-ventricular or intra-ventricular conduction deficits can be a causative factor in heart failure, which refers to a clinical syndrome in which an abnormality of cardiac function causes cardiac output to fall below a level adequate to meet the metabolic demand of peripheral tissues. In order to treat these problems, implantable cardiac devices have been developed that provide appropriately timed electrical stimulation to one or more heart chambers in an attempt to improve the coordination of atrial and/or ventricular contractions, termed cardiac resynchronization therapy (CRT). Ventricular resynchronization is useful in treating heart failure because, although not directly inotropic, resynchronization can result in a more coordinated contraction of the ventricles with improved pumping efficiency and increased cardiac output. Currently, a most common form of CRT applies stimulation pulses to both ventricles, either simultaneously or separated by a specified biventricular offset interval, and after a specified atrio-ventricular delay interval with respect to the detection an intrinsic atrial contraction.

Cardiac pacing therapy, if delivered synchronously, is demand-based. That is, pacing pulses are delivered only when the heart's intrinsic rhythm fails to maintain an adequate heart rate. Cardiac electro-stimulation delivered for purposes other than to enforce a minimum rate, however, is currently delivered in a more or less constant manner without regard for changes in the patient's condition.

Summary

The present invention relates to a method and device for delivering cardiac function therapy on a demand basis. In accordance with the invention, an implantable device for delivering cardiac function therapy is programmed to suspend such therapy at periodic intervals or upon command from an external programmer. Measurements related to hemodynamic performance are then taken using one or more sensing modalities incorporated into the device. Based upon these measurements, the device

uses a decision algorithm to determine whether further delivery of the cardiac function therapy is warranted.

Brief Description of the Drawings

Fig. 1 is a system diagram of a cardiac device configured for multi-site stimulation and sensing.

Fig. 2 is a block diagram of exemplary components for computing the LF/HF ratio.

Fig. 3 illustrates an exemplary algorithm for implementing the invention.

Detailed Description

As noted above, most current cardiac pacing devices are demand based, that is, any pacing mode or pacemaker that delivers an output pulse only when the intrinsic rate is less than the programmed base rate. Thus, a demand interval specifies the time period between two consecutive paced events in the same chamber without an intervening sensed event. As described below, implantable devices for delivering cardiac function therapies may be prescribed for post-MI patients or heart failure patients in order to boost cardiac output and/or to reverse cardiac remodeling. In such cases, the cardiac function therapy delivery can be made available on a demand basis in accordance with the present invention.

1. Cardiac Function Therapy

One example of electro-stimulatory therapy for the purpose of improving cardiac function is CRT. In ventricular resynchronization therapy, the ventricles are paced at more than one site in order to affect a spread of excitation that results in a more coordinated contraction and thereby overcome interventricular or intraventricular conduction defects. Biventricular pacing is one example of resynchronization therapy in which both ventricles are paced in order to synchronize their respective contractions. Resynchronization therapy may also involve multi-site pacing applied to only one chamber. For example, a ventricle may be paced at multiple sites with

excitatory stimulation pulses in order to produce multiple waves of depolarization that emanate from the pacing sites. This may produce a more coordinated contraction of the ventricle and thereby compensate for intraventricular conduction defects that may exist.

Another type of cardiac function therapy is stress reduction pacing which involves altering the coordination of ventricular contractions with multi-site pacing in order to change the distribution of wall stress experienced by the ventricle during the cardiac pumping cycle. The degree to which a heart muscle fiber is stretched before it contracts is termed the preload. The maximum tension and velocity of shortening of a muscle fiber increases with increasing preload. The increase in contractile response of the heart with increasing preload is known as the Frank-Starling principle. When a myocardial region contracts late relative to other regions, the contraction of those opposing regions stretches the later contracting region and increases the preload. The degree of tension or stress on a heart muscle fiber as it contracts is termed the afterload. Because pressure within the ventricles rises rapidly from a diastolic to a systolic value as blood is pumped out into the aorta and pulmonary arteries, the part of the ventricle that first contracts due to an excitatory stimulation pulse does so against a lower afterload than does a part of the ventricle contracting later. Thus a myocardial region that contracts later than other regions is subjected to both an increased preload and afterload. This situation is created frequently by the ventricular conduction delays associated with heart failure and ventricular dysfunction. The heart's initial physiological response to the uneven stress resulting from an increased preload and afterload is compensatory hypertrophy in those later contracting regions of the myocardium. In the later stages of remodeling, the regions may undergo atrophic changes with wall thinning due to the increased stress. The parts of the myocardium that contract earlier in the cycle, on the other hand, are subjected to less stress and are less likely to undergo hypertrophic remodeling. This phenomena may be used to effect reversal of remodeling by pacing one or more sites in a ventricle (or an atrium) with one or more excitatory stimulation pulses during a cardiac cycle with a specified pulse output sequence. The pace or paces are delivered in a manner that excites a previously

stressed and remodeled region of the myocardium earlier during systole so that it experiences less afterload and preload. This pre-excitation of the remodeled region relative to other regions unloads the region from mechanical stress and allows reversal of remodeling to occur.

2. Hardware platform

An implantable cardiac device is typically placed subcutaneously or submuscularly in a patient's chest with leads threaded intravenously into the heart to connect the device to electrodes used for sensing and stimulation. Leads may also be positioned on the epicardium by various means. A programmable electronic controller causes the stimulus pulses to be output in response to lapsed time intervals and sensed electrical activity (i.e., intrinsic heart beats not as a result of a stimulus pulse). The device senses intrinsic cardiac electrical activity by means of internal electrodes disposed near the chamber to be sensed. A depolarization wave associated with an intrinsic contraction of the atria or ventricles that is detected by the device is referred to as an atrial sense or ventricular sense, respectively. In order to cause such a contraction in the absence of an intrinsic beat, a stimulus pulse (a.k.a. a pace or pacing pulse when delivered in order to enforce a certain rhythm) with energy above a certain threshold is delivered to the chamber.

Fig. 1 shows a system diagram of a microprocessor-based cardiac device suitable for practicing the present invention. The device is equipped with multiple sensing and pacing channels which may be physically configured to sense and/or pace multiple sites in the atria or the ventricles. The device shown in Fig. 1 can be configured for cardiac resynchronization pacing of the atria or ventricles and/or for myocardial stress reduction pacing such that one or more cardiac sites are sensed and/or paced in a manner that pre-excites at least one region of the myocardium. The multiple sensing/stimulation channels may be configured, for example, with one atrial and two ventricular sensing/stimulation channels for delivering biventricular resynchronization therapy, with the atrial sensing/stimulation channel used to deliver biventricular resynchronization therapy in an atrial tracking mode as well as to pace

the atria if required. The controller 10 of the pacemaker is a microprocessor which communicates with a memory 12 via a bidirectional data bus. The memory 12 typically comprises a ROM (read-only memory) for program storage and a RAM (random-access memory) for data storage. The controller could be implemented by other types of logic circuitry (e.g., discrete components or programmable logic arrays) using a state machine type of design, but a microprocessor-based system is preferable. As used herein, the term "circuitry" should be taken to refer to either discrete logic circuitry or to the programming of a microprocessor.

Shown in the figure are four exemplary sensing and pacing channels designated "a" through "d" comprising bipolar leads with ring electrodes 34a-d and tip electrodes 33a-d, sensing amplifiers 31a-d, pulse generators 32a-d, and channel interfaces 30a-d. Each channel thus includes a pacing channel made up of the pulse generator connected to the electrode and a sensing channel made up of the sense amplifier connected to the electrode. The channel interfaces 30a-d communicate bidirectionally with microprocessor 10, and each interface may include analog-to-digital converters for digitizing sensing signal inputs from the sensing amplifiers and registers that can be written to by the microprocessor in order to output pacing pulses, change the pacing pulse amplitude, and adjust the gain and threshold values for the sensing amplifiers. The sensing circuitry of the pacemaker detects a chamber sense, either an atrial sense or ventricular sense, when an electrogram signal (i.e., a voltage sensed by an electrode representing cardiac electrical activity) generated by a particular channel exceeds a specified detection threshold. Pacing algorithms used in particular pacing modes employ such senses to trigger or inhibit pacing, and the intrinsic atrial and/or ventricular rates can be detected by measuring the time intervals between atrial and ventricular senses, respectively.

The electrodes of each bipolar lead are connected via conductors within the lead to a MOS switching network 70 controlled by the microprocessor. The switching network is used to switch the electrodes to the input of a sense amplifier in order to detect intrinsic cardiac activity and to the output of a pulse generator in order to deliver a pacing pulse. The switching network also enables the device to sense or pace

either in a bipolar mode using both the ring and tip electrodes of a lead or in a unipolar mode using only one of the electrodes of the lead with the device housing or can 60 serving as a ground electrode. As explained below, one way in which the device may alter the spatial distribution of pacing is to switch from unipolar to bipolar pacing (or vice-versa) or to interchange which electrodes of a bipolar lead are the cathode and anode during bipolar pacing. A shock pulse generator 50 is also interfaced to the controller for delivering a defibrillation shock via a pair of shock electrodes 51 to the atria or ventricles upon detection of a shockable tachyarrhythmia.

The controller 10 controls the overall operation of the device in accordance with programmed instructions stored in memory, including controlling the delivery of paces via the pacing channels, interpreting sense signals received from the sensing channels, and implementing timers for defining escape intervals and sensory refractory periods. An exertion level sensor 330 (e.g., an accelerometer, a minute ventilation sensor, or other sensor that measures a parameter related to metabolic demand) enables the controller to adapt the pacing rate in accordance with changes in the patient's physical activity. A telemetry interface 40 is also provided which enables the controller to communicate with an external programmer.

In one embodiment, the exertion level sensor is a minute ventilation sensor which includes an exciter and an impedance measuring circuit. The exciter supplies excitation current of a specified amplitude (e.g., as a pulse waveform with constant amplitude) to excitation electrodes that are disposed in the thorax. Voltage sense electrodes are disposed in a selected region of the thorax so that the potential difference between the electrodes while excitation current is supplied is representative of the transthoracic impedance between the voltage sense electrodes. The conductive housing or can may be used as one of the voltage sense electrodes. The impedance measuring circuitry processes the voltage sense signal from the voltage sense electrodes to derive the impedance signal. Further processing of the impedance signal allows the derivation of signal representing respiratory activity and/or cardiac blood volume, depending upon the location the voltage sense electrodes in the thorax. (See, e.g., U.S. Patent Nos. 5,190,035 and 6,161,042, assigned to the assignee of the present

invention and hereby incorporated by reference.) If the impedance signal is filtered to remove the respiratory component, the result is a signal that is representative of blood volume in the heart at any point in time, thus allowing the computation of stroke volume and, when combined with heart rate, computation of cardiac output.

The controller is capable of operating the device in a number of programmed pacing modes which define how pulses are output in response to sensed events and expiration of time intervals. Most pacemakers for treating bradycardia are programmed to operate synchronously in a so-called demand mode where sensed cardiac events occurring within a defined interval either trigger or inhibit a pacing pulse. Inhibited demand pacing modes utilize escape intervals to control pacing in accordance with sensed intrinsic activity such that a pacing pulse is delivered to a heart chamber during a cardiac cycle only after expiration of a defined escape interval during which no intrinsic beat by the chamber is detected. Escape intervals for ventricular pacing can be restarted by ventricular or atrial events, the latter allowing the pacing to track intrinsic atrial beats. Cardiac function therapy, whether for the purpose of cardiac resynchronization or for reversal of remodeling, is most conveniently delivered in conjunction with a bradycardia pacing mode where, for example, multiple excitatory stimulation pulses are delivered to multiple sites during a cardiac cycle in order to both pace the heart in accordance with a bradycardia mode and provide pre-excitation of selected sites.

A particular pacing mode for delivering cardiac function therapy, whether for stress reduction or resynchronization, includes a defined pulse output configuration and pulse output sequence, where the pulse output configuration specifies a specific subset of the available electrodes to be used for delivering pacing pulses and the pulse output sequence specifies the timing relations between the pulses. The pulse output configuration is defined by the controller selecting particular pacing channels for use in outputting pacing pulses and by selecting particular electrodes for use by the channel with switch matrix 70. The pulse output configuration and sequence which optimally effects reverse remodeling by selectively reducing myocardial wall stress may or may not be the optimum pulse output configuration and sequence for

maximizing hemodynamic performance by resynchronizing ventricular contractions. For example, a more hemodynamically effective contraction may be obtained by exciting all areas of the myocardium simultaneously, which may not effectively promote reversal of the hypertrophy or remodeling.

3. Demand-based cardiac function therapy

In order to deliver cardiac function therapy on a demand basis, the controller of an implantable cardiac device is programmed to suspend delivery of the cardiac function therapy and assess the patient's cardiac function by means of one or more sensing modalities. A decision algorithm is then employed to determine subsequent therapy. In one embodiment, a binary decision algorithm either continues or indefinitely suspends the cardiac function therapy based upon the cardiac function assessment. For example, stress reduction therapy may be employed in a heart failure or post-MI patient to effect reversal of cardiac remodeling. If the cardiac function assessment indicates that the patient's condition is unchanged or has deteriorated, the device resumes the stress reduction therapy. If the patient's cardiac function has improved sufficiently, on the other hand, the device indefinitely terminates the therapy. Delivery of cardiac resynchronization therapy may similarly be continued or terminated based upon the cardiac function assessment. After termination of cardiac function therapy, the device may continue to monitor the patient's cardiac function, periodically or otherwise, so that therapy can be resumed if needed. In addition, the device may periodically switch on therapies and see whether patient conditions have evolved gradually and therapy delivery is warranted again.

In another embodiment, the delivery of cardiac function therapy is modified in accordance with the assessment of cardiac function. The cardiac function therapy may be modified by changing the pulse output configuration, the pulse output sequence, and/or various pacing parameters. For example, the device may change from a pulse output configuration and sequence considered optimal for reversal of remodeling to one considered optimal for resynchronization pacing or vice-versa as a result of the cardiac function assessment. In another example, pacing parameters such

as the length of one or more escape intervals, a biventricular offset interval for biventricular pacing, or an AV delay interval for atrial tracking or AV sequential pacing are changed in accordance with the cardiac function assessment.

a. Assessment of cardiac function

One means by which cardiac function may be assessed is by measuring cardiac output and comparing it with the patient's measured exertion level. As described earlier, cardiac output may be measured by an impedance technique in which transthoracic impedance is measured and used compute stroke volume. The stroke volume integrated over time (or averaged and multiplied by heart rate) gives the patient's cardiac output. A look-up table or linear function may be used to compute what cardiac output is considered adequate for a given exertion level. Based upon these measurements, the device may then decide whether cardiac function therapy is warranted.

Another means for assessing cardiac function is by determining the autonomic balance of the patient. It is well-known that an increase in the activity of the sympathetic nervous system may be indicative of metabolic stress and the need for increased cardiac output. One means by which increased sympathetic activity may be detected is via spectral analysis of heart rate variability. Heart rate variability refers to the variability of the time intervals between successive heart beats during a sinus rhythm and is primarily due to the interaction between the sympathetic and parasympathetic arms of the autonomic nervous system. Spectral analysis of heart rate variability involves decomposing a signal representing successive beat-to-beat intervals into separate components representing the amplitude of the signal at different oscillation frequencies. It has been found that the amount of signal power in a low frequency (LF) band ranging from 0.04 to 0.15 Hz is influenced by the levels of activity of both the sympathetic and parasympathetic nervous systems, while the amount of signal power in a high frequency band (HF) ranging from 0.15 to 0.40 Hz is primarily a function of parasympathetic activity. The ratio of the signal powers, designated as the LF/HF ratio, is thus a good indicator of the state of autonomic

balance, with a high LF/HF ratio indicating increased sympathetic activity. An LF/HF ratio which exceeds a specified threshold value may be taken as an indicator that cardiac function is not adequate.

A cardiac rhythm management device can be programmed to determine the LF/HF ratio by analyzing data received from its ventricular sensing channels. The intervals between successive ventricular senses, referred to as RR intervals, can be measured and collected for a period of time or a specified number of beats. In order to derive a signal representing heart rate variability during a sinus rhythm, ectopic ventricular beats (i.e., premature ventricular contractions or PVCs) can be detected by monitoring whether a P wave precedes each R wave, with the RR intervals before and after the PVC changed to an interpolated or otherwise filtered value. The resulting series of RR interval values is then stored as a discrete signal. The signal can be used directly as indexed by heartbeat such that each value of the signal represents an RR interval for a particular heartbeat. Preferably, however, the signal is resampled at a specified sampling frequency in order to equalize the time intervals between signal values and thus convert the signal into a discrete time signal, where the sampling frequency is selected to meet the Nyquist criterion with respect to the frequencies of interest. In any case, the RR interval signal can then be analyzed to determine its energies in the high and low frequency bands as described above.

Spectral analysis of an RR interval signal can be performed directly in the frequency domain using discrete Fourier transform or autoregression techniques. Frequency domain analysis is computationally intensive, however, and may not be practical in an implantable device. A time-domain technique for determining the high and low frequency components of the signal is therefore preferably used. Fig. 2 illustrates the functional components of an exemplary system for doing this that can be implemented as code executed by the controller and/or dedicated hardware components. The RR interval signal obtained as described above is input to both a low band digital filter 201 and a high band digital filter 202. The low band filter 201 is a bandpass filter with a passband corresponding to the LF band (e.g., 0.04 to 0.15 Hz), while the high band filter 202 is a bandpass filter with a passband corresponding

to the HF band (e.g., 0.15 to 0.40 Hz). The outputs of filters 201 and 202 are then input to power detectors 203 and 204, respectively, in order to derive signals proportional to the power of the RR interval signal in each of the LF and HF bands. Power detection may be performed, for example, by squaring the amplitude of the signal and integrating over a specified average time. The output of power detector 203 is thus a signal P1 that represents the power of the RR interval signal in the LF band, and the output of power detector 204 is a signal P2 representing the power in the HF band. The signals P1 and P2 are next input to a divider 205 that computes the quantity S1/S2 which equals the LF/HF ratio. The LF/HF ratio is then input to a moving average filter 206 that computes an average value for the ratio over a specified period (e.g., 5 minutes). An updated LF/HF ratio may be computed in this manner on a beat-to-beat basis.

In the above description, heart rate variability was derived from the RR interval signal during normal sinus rhythm. It should also be appreciated that, if normal sinus rhythm is present, the RR interval is equivalent to the interval between successive atrial senses. As used herein, therefore, the term RR interval should be regarded as the interval between heart beats during sinus rhythm whether the beats are atrial or ventricular. Also, as an alternative to time-domain filtering, a statistical method of estimating the LF/HF ratio may be employed as described in U.S. Patent Application Ser. No. 10/436,876 filed May 12, 2003 and herein incorporated by reference.

Other means of assessing cardiac function may also be employed to deliver demand-based cardiac function therapy. The impedance technique for measuring cardiac output discussed above may also be used to measure ventricular volumes at various stages of the cardiac cycle such as end-diastolic and end-systolic volumes and used to compute parameters reflective of cardiac function such as ejection fraction. The implantable device may also be equipped with other sensing modalities such as a pressure transducer 85 shown in Fig. 1. Such a pressure transducer may be attached to an intravascular lead and be appropriately disposed for measuring diastolic filling pressures and/or systolic pulse pressures.

b. Exemplary algorithm

Fig. 3 illustrates an exemplary algorithm for demand-based cardiac function therapy as could be implemented by appropriate programming of the implantable device controller. Starting at step 200, the device initiates cardiac function therapy such as stress reduction pacing or resynchronization pacing with a specified pacing mode and using a specified pulse output configuration and pulse output sequence. At step 201, either at periodic intervals or upon command from an external programmer via the telemetry interface, the device suspends further delivery of cardiac function therapy. At step 202, the device next begins an assessment of the patient's cardiac function using one or more sensing modalities. In this embodiment, the device computes cardiac output by measurement of the heart rate and cardiac stroke volume via the intra-thoracic impedance method. The patient's exertion level 203 (e.g., either activity level or minute ventilation) is then measured and compared with the measured cardiac output measurement at step 203 to determine whether the patient's cardiac output is adequate for that particular exertion level. Based upon this comparison a first cardiac function parameter CFP1 may be computed at step 204 which, if below a specified threshold level, indicates inadequate cardiac function. Next, the patient's autonomic balance is assessed at step 205, and a second cardiac function parameter CFP2 is computed which is indicative of the extent of metabolic stress experienced by the patient. This parameter can also be compared with a specified threshold value for decision-making purposes. Based upon the computed cardiac function parameters CFP1 and CFP2, the device at step 206 decides whether the patient's cardiac function is inadequate. If so, the device continues cardiac function therapy by returning to step 200. If the computed cardiac function parameters indicate that the patient's cardiac function has improved to a sufficient extent, on the other hand, the device indefinitely suspends further delivery of cardiac function therapy at step 207.

Although the invention has been described in conjunction with the foregoing specific embodiments, many alternatives, variations, and modifications will be apparent to those of ordinary skill in the art. Such alternatives, variations, and modifications are intended to fall within the scope of the following appended claims.